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# Estimating 3D joint kinematics from video sequences of running and cutting maneuvers—assessing the accuracy of simple visual inspection

Tron Krosshaug [a,*], Atsuo Nakamae [a], Barry Boden [b], Lars Engebretsen [a], Gerald Smith [c], James Slauterbeck [a,d], Timothy E. Hewett [e,f,g], Roald Bahr [a]

[a] Oslo Sports Trauma Research Center, Department of sports Medicine, Norwegian School of Sport Sciences, Oslo, Norway
[b] The Orthopaedic Center, Rockville, MD, USA
[c] Department for physical performance, Norwegian School of Sport Sciences, Oslo, Norway
[d] University of Vermont, Department of Orthopaedics & Rehabilitation, Burlington, VT, USA
[e] Cincinnati Children's Hospital Research Foundation, Sports Medicine Biodynamics Center and Human Performance Laboratory, Cincinnati, OH, USA
[f] University of Cincinnati College of Medicine, Departments of Pediatrics and Orthopaedic Surgery, Cincinnati, OH, USA
[g] University of Cincinnati College of Medicine, Departments of Rehabilitation Sciences and Bioengineering, Cincinnati, OH, USA

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## Abstract

Videos of sports injuries can potentially provide valuable information on non-contact ACL injuries. However, only the un-validated simple visual inspection approach has, so far, been used. Therefore, the purpose of this study was to test the accuracy and precision of researchers in estimating kinematics from video sequences of situations that typically lead to ACL injuries. We also tested if accuracy and precision could be improved through a training program.

Using a routine surface marker based infrared, 240 Hz, 3D motion analysis system, we recorded running and cutting trials from three test subjects. Six observers were asked to provide estimates of kinematic variables from 27 video composites from one, two or three ordinary cameras, systematically varying viewing angles and time point of analysis. The observers thereafter went through a training program where 35 similar composites were analyzed, and feedback on the kinematics, as measured by the 3D motion analysis system, was provided on a group basis. Finally, the test was repeated to assess accuracy and precision.

The mean error for knee flexion was $-19^\circ$, indicating a consistent underestimation. Hip flexion was underestimated by $7^\circ$, but the standard deviation between the observers was $18^\circ$ on average, indicating poor consistency. Substantial errors were also found in the accuracy and precision of the other estimates. Only small group effects were seen from our training program.

Based on these findings, results from studies using a simple visual inspection approach to describe joint motion must be interpreted with caution.


*Keywords:* Athletic injuries; Anterior cruciate ligament; Biomechanics; Perception

## 1. Introduction

If sports injuries are to be prevented, understanding their mechanisms is a key factor [1]. In contrast to other research approaches used to study injury mechanisms, videotapes represent an objective source of kinematic information from actual injury situations [2]. The analysis of injury videos is therefore a potentially valuable research tool, which may provide detailed information on the mechanisms of specific sports injuries [2].

Our current understanding of the mechanisms of non-contact ACL (anterior cruciate ligament) injuries originates in large part from the analysis of videotapes of

* Corresponding author at: Oslo Sports Trauma Research Center, Norwegian School of Sport Sciences, PO Box 4014 Ullevaal Stadion, 0806 Oslo, Norway. Tel.: +47 23262000; fax: +47 23262307.
*E-mail address:* tron.krosshaug@nih.no (T. Krosshaug).

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injuries [3–6]. A consistent finding across the four studies available is that the knee was relatively straight at the time of injury [3–6]. However, in other aspects, the authors' descriptions of the injury mechanism differ. Boden et al. [5] reported that the amount of internal and external rotation at the time of injury was minimal, whereas Olsen et al. [3], Ebstrup et al. [6] and Teitz et al. [4] emphasized the role of internal/external rotation, as well as valgus stress.

In these studies a visual observation approach was used, where the joint angles and other kinematic variables were estimated simply from watching videotapes of ACL ruptures, without employing any measurement tools. Although our group has recently developed a more sophisticated method to estimate joint kinetics from video sequences [7], this is too time-consuming to apply on a large number of injury tapes. Simple visual assessment of joint kinematics may therefore be an alternative if the number of videos to be analyzed is large. However, the accuracy and precision of such analyses are unknown. Olsen et al. [3] performed interobserver tests in their study, showing that the reliability was relatively good, i.e. the average inter-observer difference was $10^\circ$ or less for all knee joint angle estimates. However, this finding does not necessarily indicate accuracy as systematic error would still be possible.

The purpose of this study was to test the accuracy and precision of researchers in estimating kinematics from video sequences of situations resembling those typically leading to ACL injuries. We also tested if accuracy and precision could be improved by a training program.

## 2. Materials and methods

Six experienced observers, experienced in ACL injury research and visual video analysis, participated in this study. To assess the accuracy of their estimates, they were asked to analyze 27 video sequences of running and side step cutting maneuvers performed in a laboratory. Their estimates were compared with the results from a conventional, marker-based motion analysis system used as the gold standard. To assess the potential for improved accuracy they then underwent a training program. First, they analyzed approximately 50 video sequences of similar situations while they were given intermittent feedback on how their estimates compared with the marker-based motion analysis system. Finally, they repeated their analysis of the initial 27 video sequences to see if their accuracy improved.

### *2.1. Laboratory trials*

Three test subjects, 25, 23 and 22 years old, performed trials of running and side step cutting maneuvers in the laboratory. In the side step maneuvers, the subject cut from left to right or right to left with weight bearing on both legs (two-leg cutting) or exclusively on one leg (one-leg cutting) (Fig. 1). We recorded the trials with three ordinary video cameras (25 Hz) placed on tripods or fixed to the wall. One camera filmed from the left rear, one directly from the right side at mid-stance, and one from the front, approximately $8^\circ$ off the laboratory anterior–posterior axis. Various combinations of S-VHS (Blaupunkt CC695, Hildesheim, Germany) and digital cameras (Sony TRV900E, Tokyo, Japan, Panasonic AG-EZ35 and MX8, Kadoma City, Osaka, Japan) were employed in the analysis.

### *2.2. The marker based motion analysis*

In addition to the three ordinary video camcorders, a seven-camera infra-red, motion analysis system (ProReflex, Qualisys Inc., Gothenburg, Sweden), and two force platforms (AMTI LG6-4-1, Watertown, MA, USA) were used simultaneously to record the motion at 240 and 960 Hz, respectively. Thirty-three reflective markers (9 mm radius) were attached to the subjects at the following landmarks: base of the fifth metatarsal, heel, lateral malleolus, tibial tuberosity, lateral femoral condyle, greater trochanter, the anterior superior iliac spine, mid-posterior superior iliac spine, seventh cervical, 10th thoracic, acromion, lateral epicondyle of the humerus and ulnar styloid process. In addition, two extra markers were placed on each shank and three extra markers on each thigh. For two of the test subjects, the tibial tuberosity, greater trochanter and the extra shank and thigh markers were replaced by strapped-on anatomically shaped rigid plates, each with four reflective markers. All markers were attached by the same examiner.

Signal processing, segment orientation, joint centers and inertia parameters were calculated according to Krosshaug and Bahr [7].

We used the Grood and Suntay [8] convention for calculating joint angles from the marker-based motion analysis (the valgus angle from these calculations is here termed "3D valgus"). We also calculated the frontal plane projected valgus angle, as described by McLean et al. ("2D valgus") [9]. Two-dimensional valgus was calculated as the angle between the hip–knee and knee–ankle vectors projected in the pelvic frontal plane. Foot–pelvis rotation was calculated as the angle between the foot antero-posterior axis and the pelvis antero-posterior axis, projected into the global transverse plane (the floor).

To determine joint movement towards flexion or extension we chose an 80 ms period, corresponding to two video frames prior to and two frames after each of the pre-defined time points in the videos (see below). We chose two different criteria to determine if the joint motion tendency should be considered positive/negative instead of neutral: a positive (or negative) change of at least $2^\circ$ both during the preceding and the following 40 ms, or a net change of at least $8^\circ$ during the 80 ms period in question. The last criterion was used to capture situations where joint motion went from neutral to a strong positive/negative change (e.g. knee flexion at initial contact).

### *2.3. The video analysis*

We generated 28 video recordings for the tests from 12 different laboratory trials of test subjects 1 and 2. The videos were generated by combining different cameras views systematically: front only, side only, rear only, front/side, front/rear, side/rear, and front/side/rear (Fig. 1). The composite videos, which included more than one camera recording, were synchronized manually by using the initial contact of the foot in each camera view as the key frame. We produced four different cases for each of the seven camera view combinations, systematically changing the test subject and cutting direction. However, due to a

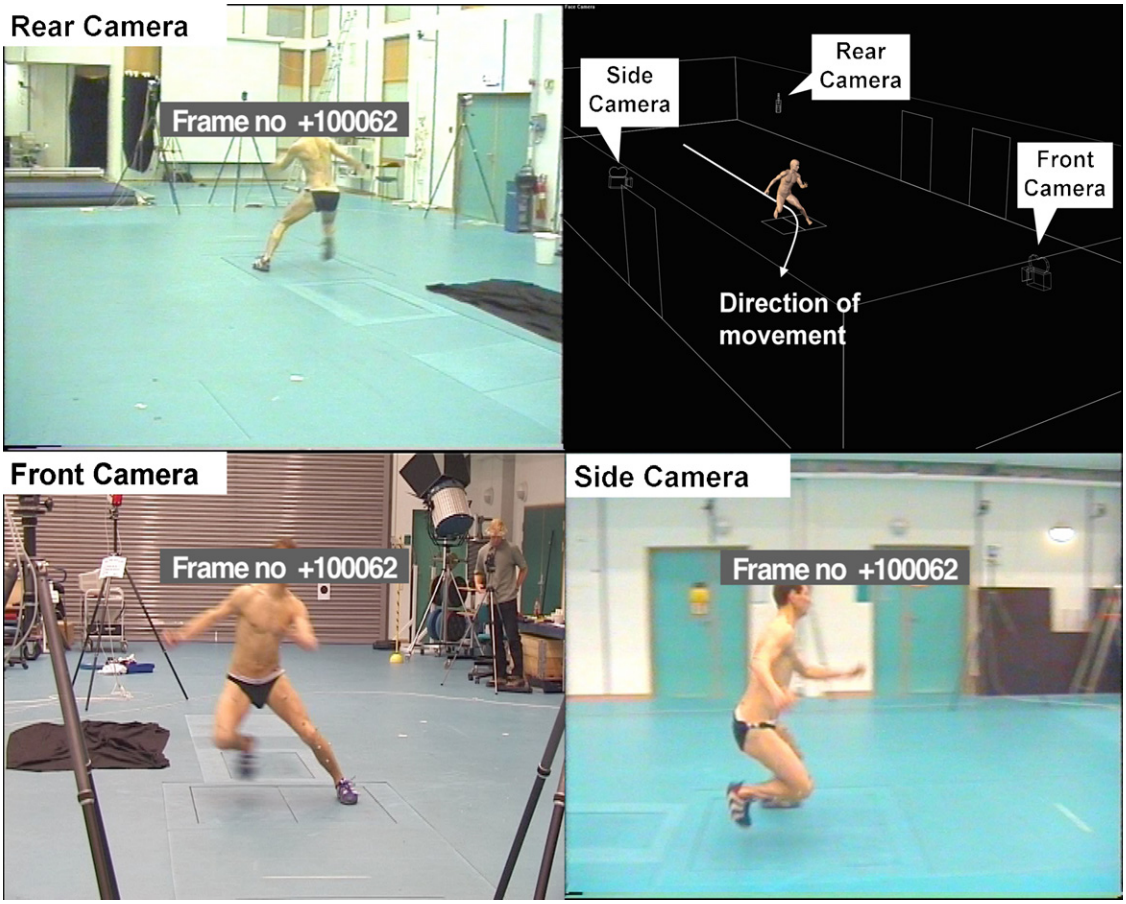


Fig. 1. Example of the synchronized video images seen in a three-camera view. The upper right panel shows an overview of the camera placement in relation to the test subject.

video editing error, one of the front/rear composites could not be used, which resulted in a total number of 27 videos available for the tests. In each video, a key frame was marked at pre-defined time points, either initial contact (IC), 100 ms after IC or 200 ms after IC. The time point was also systematically changed between each of the cases.

The video recordings were processed using Final Cut Pro HD (Version 4.5, Apple Cupertino, CA, USA), deinterlaced and stored using either the DV or the DVCPRO50 codec in PAL format. Each video was composed in two versions, one in real time and one in slow motion (50% of normal speed). Each of the analysts used a Macintosh computer with a 20″ or 21″ LCD monitor, and the analysis was performed independently, blinded to the results of the other observers. QuickTime (Version 7.0) was used to play the videos, and the observers could move the video sequence back and forth frame-by-frame using the keyboard arrows.

The observers were asked to provide estimates for the pre-defined frames marked in each of the videos by assessing knee flexion/extension, knee varus/valgus, knee internal/external rotation, hip flexion/extension, hip adduction/abduction, hip internal/external rotation, approach velocity, vertical velocity, cutting angle, and internal/external rotation of the foot relative to the pelvis using a standard form. In addition, they were asked to assess the direction of motion for the knee and hip joint angles at the pre-defined time points.

### 2.4. Training session and post-training assessment

Thirty-five videos from five cases of sidestep cutting, of which four cases were based on test subject 3 and one case was based on a new recording of test subject 2, were produced for the training session, systematically varying different camera view combinations and cutting directions in the same way as for the test videos. The training session started by first using the single camera videos, then multiple camera view composites were analyzed. All three time points were analyzed and the correct answers (i.e. the results from marker-based motion analysis) were provided at regular intervals with group discussions. During the training session discussions it became evident that the observers had used somewhat different approaches to estimate valgus angles. It was therefore decided that in the repeat-test, both the standard 3D valgus angle of Grood and Suntay [8], as well as the 2D valgus angle as suggested by McLean et al. [9] should be reported. After the training session had been completed, the observers reanalyzed the same 27 cases as before to establish whether the training had any effect on accuracy.

### 2.5. Data reporting and statistical methods

All calculations were performed using Matlab (Version 6.1, The Mathworks Inc., Natick, MA, USA). Descriptive statistics were calculated using SPSS (Version 13, SPSS Inc., Chicago, IL,

Table 1
Mean error (° or m/s) and standard deviations for the mean error (S.D.) with range for the pre- and post-tests

| | Pre-test | N | Post-test | N | Pre- to post-difference | N | p mean | p S.D. |
|---|---|---|---|---|---|---|---|---|
| Knee flexion (°) | −19 ± 14 [−49,31] | 160 | −18 ± 15 [−54,21] | 154 | 1[−1 to 3] | 152 | 0.47 | 0.30 |
| 2D valgus (°) | 4 ± 9 [−12,35] | 158 | 5 ± 10 [−15,35] | 150 | 1[0 to 2] | 149 | 0.005 | 0.12 |
| 3D valgus (°) | −2 ± 6 [−19,16] | 158 | −5 ± 6 [−19,11] | 152 | −3[−4 to −2] | 151 | 0.000 | 0.01 |
| Knee internal rotation (°) | −12 ± 11 [−46,11] | 162 | −9 ± 8 [−29,8] | 154 | 3[2 to 5] | 154 | 0.000 | 0.000 |
| Hip flexion (°) | −7 ± 18 [−50,48] | 162 | −7 ± 19 [−50,43] | 159 | 0[−2 to 2] | 159 | 0.93 | 0.40 |
| Hip abduction (°) | 4 ± 10 [−27,34] | 158 | 2 ± 13 [−35,37] | 149 | −2 [−4 to 0] | 145 | 0.046 | 0.53 |
| Hip internal rotation (°) | −10 ± 16 [−57,17] | 160 | −10 ± 14 [−37,13] | 154 | 0 [−2 to 2] | 152 | 0.85 | 0.16 |
| Approach speed (m/s) | 0.1 ± 1.1 [−3.1,2.6] | 135 | 0.3 ± 0.9 [−2.1,3.4] | 135 | −0.2 [−0.4 to 0.0] | 135 | 0.02 | 0.053 |
| Vertical speed (m/s) | −0.5 ± 1.0 [−2.4,4.1] | 123 | −0.5 ± 1.0 [−2.7,2.4] | 123 | 0.0 [−0.2 to 0.2] | 123 | 0.8 | 0.58 |
| Cutting angle (°) | 2 ± 19 [−41,47] | 159 | 2 ± 19 [−51,39] | 162 | 1 [−1 to 3] | 159 | 0.36 | 0.07 |
| Foot-pelvis rotation (°) | 1 ± 11 [−31,28] | 160 | 0 ± 14 [−33,43] | 162 | −1 [−2 to 0] | 160 | 0.095 | 0.002 |

The mean difference between the pre- and post-test is also shown, with the corresponding 95% confidence interval (95% CI). *P*-values are reported for pre- to post-test changes in the mean error (*p* mean) and the standard deviation of the mean error (*p* S.D.).

USA). Knee flexion, abduction and internal rotation angles as well as hip flexion, abduction and internal rotation are shown as positive values. Approach speed was the instantaneous horizontal velocity of the center of mass at IC. Similarly, vertical speed was the downward directed velocity of the center of mass at IC.

We calculated the difference between each of the analysts' estimates and the marker-based measurements as the gold standard for each variable in the first test. The accuracy of the estimates was reported as the mean difference from the gold standard. The precision of the estimates was reported as the standard deviation (S.D.) of this difference between the observers, averaged over cases, (with range). We used a paired *t*-test to examine if the training led to significant improvements in the means (accuracy) for the differences between the estimates and the gold standard. When testing if the S.D.s (precision) had improved, we first calculated the absolute difference to the mean of the six estimates for each analyst, each variable and each situation for the pre and post tests. The pre-post difference was then tested with a paired *t*-test. We reported the mean difference between the pre- and post-test with 95% confidence intervals (95% CI). Through generalized estimating equation analyses, we then tested whether the following factors had a significant influence on the observed mean differences from the gold standard: The true angle/velocity, time point (IC, 100 ms, 200 ms), number of cameras (single or multiple), inclusion of a side camera (yes, no), inclusion of a front camera (yes, no), analyzed leg on same side as the side camera (yes, no), movement type (running, cutting), and subject (1, 2). This analysis was undertaken with the software package STATA8 (StataCorp LP, Natrick, TX, USA), using an unstructured correlation matrix. To determine the factors to be included in the equation, we undertook a pre-screening of the correlations between the factors. If correlations were higher than 0.6, one of the two factors was removed based on a professional judgment of what was most relevant for the studied variable.

For all analyses, an alpha level of <.05 was used to denote statistical significance. For the categorical variables (joint motion), a kappa-test was used to compare the agreement between the estimates and the gold standard. The strength of agreement can be classified as follows: poor (value: <.20), fair (.21–.40), moderate (.41–.60), good (.61–.80), and very good (.81–1.00) [10].

## 3. Results

Substantial errors were found in the accuracy (Table 1). The mean error for knee flexion was −19°, indicating a consistent underestimation. Hip angles were also underestimated systematically by an average of 7°. Both hip and knee internal rotation were underestimated by 10° and 12°, respectively. As seen by the large standard deviations and maximal errors, precision was also poor, in particular for hip flexion, hip internal rotation and knee flexion.

Relatively small overall systematic errors were seen in cutting angle and external foot rotation. However, the

Table 2
Mean error (° or m/s) and standard deviations of the mean error as a function of camera view combinations

| | Front | Side | Rear | Front/side | Front/rear | Side/rear | Front/side/rear |
|---|---|---|---|---|---|---|---|
| Knee flexion (°) | −25 ± 15 | −16 ± 9 | −16 ± 12 | −11 ± 13 | −28 ± 19 | −17 ± 14 | −20 ± 11 |
| 2D valgus (°) | 3 ± 9 | 6 ± 13 | 7 ± 12 | 4 ± 8 | 5 ± 9 | 3 ± 8 | 3 ± 7 |
| 3D valgus (°) | −2 ± 7 | 0 ± 6 | −2 ± 5 | 1 ± 6 | −5 ± 8 | −5 ± 6 | 0 ± 4 |
| Knee internal rotation (°) | −11 ± 9 | −11 ± 14 | −14 ± 13 | −5 ± 9 | −18 ± 8 | −16 ± 12 | −12 ± 6 |
| Hip flexion (°) | −14 ± 16 | −3 ± 20 | 1 ± 22 | −12 ± 8 | −5 ± 24 | −4 ± 17 | −13 ± 13 |
| Hip abduction (°) | 4 ± 11 | 1 ± 11 | 1 ± 9 | 0 ± 7 | 8 ± 13 | 11 ± 13 | 4 ± 9 |
| Hip internal rotation (°) | −5 ± 15 | −16 ± 20 | −11 ± 14 | −10 ± 13 | −7 ± 17 | −14 ± 16 | −8 ± 14 |
| Approach speed (m/s) | 0.3 ± 0.9 | −0.4 ± 1.3 | 0.2 ± 1.4 | −0.3 ± 1.0 | 0.6 ± 0.9 | 0.3 ± 0.7 | −0.2 ± 1.1 |
| Vertical speed (m/s) | −0.5 ± 1.2 | −0.5 ± 0.9 | −0.5 ± 1.0 | −0.5 ± 0.8 | −0.4 ± 1.5 | −0.6 ± 1.2 | −0.5 ± 0.7 |
| Cutting angle (°) | 4 ± 19 | 6 ± 18 | −5 ± 10 | 6 ± 15 | −3 ± 25 | 4 ± 23 | −2 ± 17 |
| Foot external rotation (°) | 1 ± 11 | −1 ± 13 | 0 ± 14 | 0 ± 10 | −1 ± 8 | 8 ± 8 | 2 ± 11 |

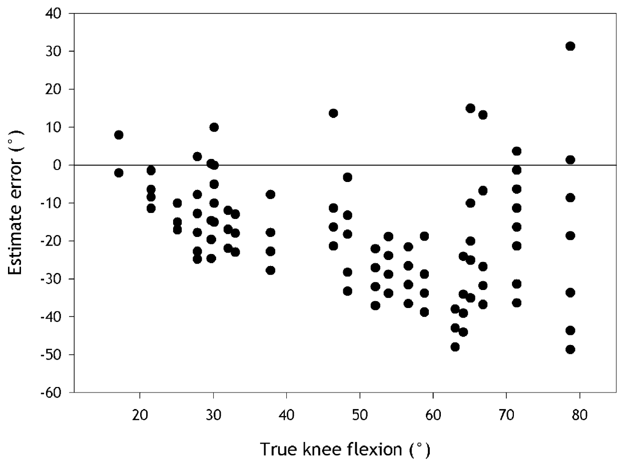


Fig. 2. Error in knee flexion angle estimates of the analysts as function of the true angle, measured with the 3D motion analysis system.

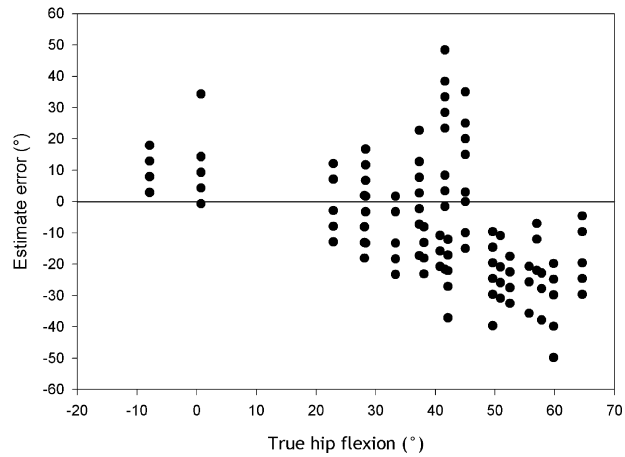


Fig. 4. Error in hip flexion angle estimates of the analysts as function of the true angle, measured with the 3D motion analysis system.

standard deviations were considerable. For approach and vertical speed small mean errors were seen, while the standard deviations were close to 1 m/s (Table 1).

Only small overall changes in the mean error and standard deviations were seen from the pre- to the post-training test, and for more than half of the variables the post-training results were no better than pre-training (Table 1). On an individual basis, however, significant changes were observed, and two of the observers reached a decreased mean error in knee flexion of more than 8°.

Table 2 shows mean errors and standard deviations using different camera combinations. From the multi-variate regression analysis, we found that the knee flexion estimate errors were significantly less (−6° versus −37°) when a side camera was present ($p = 0.02$), when the right leg (i.e. closest to the side camera) was analyzed ($p < 0.001$) and when the flexion angle was lower than 30° ($p < 0.001$). The relationship between the estimate error and true knee joint angle is displayed in Fig. 2.

Figs. 3–5 illustrate the error characteristics across analysts and cases for three other key variables. A significant relationship was observed between the mean estimate error and the true joint angle for 3D valgus ($p < 0.001$) and hip flexion ($p < 0.001$), but not for hip abduction ($p = 0.11$).

A significant relationship between the variable mean error and the true value was also found for 2D valgus ($p < 0.001$), 3D valgus ($p < 0.001$), knee rotation ($p < 0.001$), hip rotation ($p < 0.001$), approach speed ($p < 0.001$), foot-pelvis angle ($p < 0.001$) and cutting angle ($p = 0.006$).

We did not find any relationship between the mean estimate error and inclusion of a side camera for hip flexion. For hip adduction/abduction, we found a significantly lower ($p = 0.017$) mean estimate error in cases that included a front camera view compared to those that did not. The mean error estimate was also significantly better in the running trials compared with the cutting maneuvers ($p < 0.001$).

We found no significant differences in the mean estimate error for varus/valgus angles between cases that included a

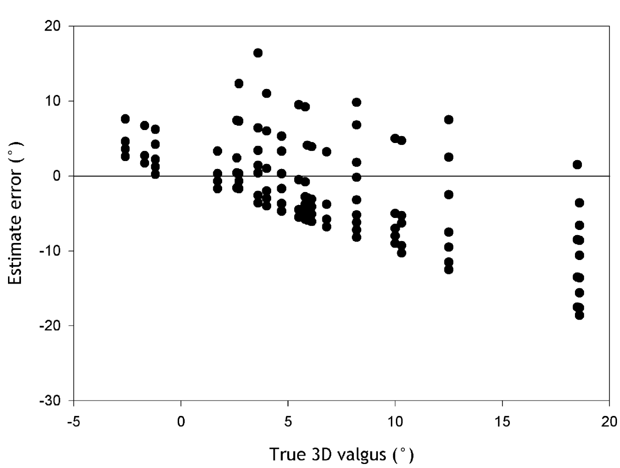


Fig. 3. Error in knee 3D valgus angle estimates of the analysts as function of the true angle, measured with the 3D motion analysis system.

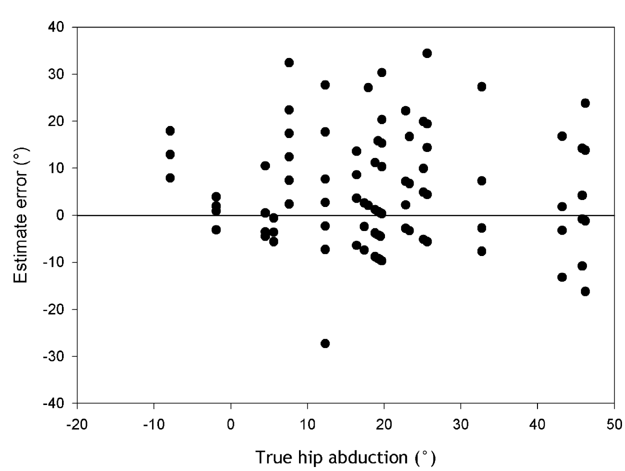


Fig. 5. Error in hip abduction angle estimates of the analysts as function of the true angle, measured with the 3D motion analysis system.

Table 3
Agreement between estimated and actual tendency of joint motion for knee flexion/extension

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Flexion | Neutral | Extension |
| Flexion | 74 | 14 | 15 |
| Neutral | 4 | 1 | 1 |
| Extension | 6 | 15 | 32 |

$N = 162$; Kappa = 0.40.

Table 4
Agreement between estimated and actual tendency of joint motion for knee varus/valgus

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Varus | Neutral | Valgus |
| Varus | 1 | 3 | 0 |
| Neutral | 3 | 45 | 15 |
| Valgus | 0 | 42 | 35 |

$N = 144$; Kappa = 0.19.

Table 5
Agreement between estimated and actual tendency of joint motion for knee internal/external rotation

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Internal rotation | Neutral | External rotation |
| Internal rotation | 4 | 22 | 2 |
| Neutral | 6 | 25 | 1 |
| External rotation | 13 | 45 | 3 |

$N = 120$; Kappa = −0.02.

Table 6
Agreement between estimated and actual tendency of joint motion for hip flexion/extension

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Flexion | Neutral | Extension |
| Flexion | 26 | 43 | 13 |
| Neutral | 2 | 7 | 0 |
| Extension | 1 | 27 | 41 |

$N = 160$; Kappa = 0.27.

Table 7
Agreement between estimated and actual tendency of joint motion for hip adduction/abduction

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Adduction | Neutral | Abduction |
| Adduction | 1 | 6 | 0 |
| Neutral | 4 | 23 | 8 |
| Abduction | 6 | 55 | 42 |

$N = 145$; Kappa = 0.11.

Table 8
Agreement between estimated and actual tendency of joint motion for hip internal/external rotation

| Estimated joint motion | True joint motion | | |
|---|---|---|---|
| | Internal rotation | Neutral | External rotation |
| Internal rotation | 9 | 19 | 3 |
| Neutral | 11 | 28 | 14 |
| External rotation | 14 | 33 | 18 |

$N = 149$; Kappa = 0.04.

front camera view and those that did not. The number of cameras (one versus two or more) did not affect the mean estimate error for any of the variables.

The overall agreement between estimated and actual direction of the joint motion was poor (Tables 3–8). The relative match was best for knee flexion/extension (Table 3), where a kappa value of 0.40 indicated a fair agreement. However, as many as 13% of the estimates were still highly biased, i.e. estimating extension when flexion was correct or vice versa. The lowest scores were obtained for internal/external rotation of the knee and hip, with kappa-values of −0.02 and 0.04 (Tables 5 and 8).

## 4. Discussion

The aim of this study was to test the accuracy and precision of researchers to estimate kinematics from video sequences of situations resembling those typically leading to ACL injuries and to examine whether or not a structured feedback training program would lead to improvements. The results clearly demonstrated that the accuracy and precision of the estimates across observers and trials was generally poor, and only small changes were seen as a result of the training session.

The substantial estimation errors indicate that previous studies using a simple visual inspection approach to describe the mechanisms for ACL injury must be interpreted with caution. In particular, the current belief regarding the knee flexion angle at the time of rupture (i.e. that the knee is relatively straight with less than 30° knee flexion) [3–6] may not be true. Our data showed that the measured knee angle was twice as high as the estimated (Fig. 2). Although the estimate improved somewhat when a side camera was present and nothing blocked the knee view, the accuracy was still poor.

As outlined by Bahr and Krosshaug, visual analysis helps with understanding of injury mechanisms and therefore may be important in the development of preventative measures [1]. However, until methods for visual analysis of injury videos improve significantly, we should consider the reported results as rough descriptions only. It is possible that such methods could be used to detect trends in analyzing large numbers of cases, where random errors can be minimized through averaging. Large systematic errors were observed, however, and more sophisticated methods [7] should be used to obtain estimates of joint kinematics with acceptable accuracy.

It is well known that several sources of error may be present using a conventional surface marker approach [11–14]. In a recent study, using intra-cortical bone-pins, Benoit et al. [13] found average rotational errors up to 13.1° during cutting tasks. Reinschmidt et al. [11] reported maximal rotational errors up to 13.3° in their analysis of running. Stagni et al. [15] suggested that bone-pin studies under-estimate the error considerably, since bone-pins themselves reduce soft tissue artifacts. Such errors influence flexion/extension angles to a much lesser degree than ad-/abduction and internal-/external rotation [11,13,16]. Two-dimensional valgus calculations, based solely on joint centers, will also be virtually unaffected by skin movement artifacts. However, even if the evaluation of the accuracy could be affected by a sub-optimal gold standard, the large between-analyst standard deviations clearly demonstrate poor ability to estimate joint kinematics using simple visual inspection.

Using foot contact to synchronize the video recordings with the marker-based motion analysis recording could result in error. However, even if the knee flexion angle could change up to 20° during a 20 ms-period, corresponding to the interval between video frames, such dramatic joint angle changes were only seen between 20 and 70 ms after IC. At IC, 100 and 200 ms after IC were most often less than 5°. The potential error due to synchronization at IC, 100 and 200 ms after IC should therefore be minimal.

The substantial systematic and random errors observed in this study suggest that it is difficult, even for experienced observers, to estimate joint kinematics from video using visual observation. Most striking was the systematic underestimation of knee flexion, especially at flexion angles higher than 30°. One reason for this could be the out-of-plane position of the leg when the hip is abducted or rotated. Furthermore, the view of a limb could be obstructed by the contralateral limb. Both problems are illustrated in Fig. 1, where the observers estimated the knee flexion angle at 27.5° on average, while it was in fact 56.6°. The knee flexion error was significantly lower near the position of full extension (see Fig. 2), due to the narrower range of possible joint angle in this position. Although hip flexion estimates were not under/overestimated as much as knee flexion, the large standard deviation of 18° suggests that it is inherently difficult to assess kinematics from video. Contrary to the knee flexion findings, we did not find significant improvements in hip flexion estimates with the addition of a side camera. The same was the case for the hip abduction results where errors were considerable at all joint angles (see Fig. 5). There could be several explanations why all the hip joint angle estimates had such large standard deviations. First, it is difficult to assess pelvic orientation and hip joint center visually. Second, it is difficult to estimate joint angles when extensive movements take place in all three planes.

Another problem is that, even when an observer's perception of the segment orientation is correct, reporting the joint angles remains a challenge: there are several possibilities to perform three rotations in order to achieve the observed joint configuration and segment orientation [17]. This may indeed represent a significant problem in the present study. For example, both adduction/abduction and rotation angles in the knee and hip differed up to 15° depending on whether the JCS method as described by Grood and Suntay [8] or the attitude angle method as described by Woltring [17] was used for the calculation.

The fact that our structured feedback program did not succeed could either indicate that the training program was inadequate, or simply that the task of segment orientation perception and joint angle calculation is inherently difficult. We chose to give structured feedback to the whole group to ''calibrate'' the observers to the measurements from the motion analysis system. It is possible that an individual program, allowing the observers more time to explore their own results would have yielded better results.

## 5. Conclusion

The accuracy of the simple visual inspection approach was poor, with considerable systematic as well as random error. Minimal group effects were seen from our training program aiming at improving accuracy. Based on these findings, results from studies using a simple visual observation approach to describe joint motion must be interpreted with caution. Such studies may provide gross descriptive information but are likely to lack accuracy and precision.

## Conflict of interest statement

No conflicts of interest exist for any of the authors.

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